Closed-Loop Control
of Functional Neuromuscular Stimulation

NIH Neuroprosthesis Program Contract Number N01-NS-6-2338
Quarterly Progress Report #11
October 1, 1998 to December 31, 1998

Investigators:
Patrick E. Crago, Ph.D.
Clayton L. Van Doren, Ph.D.
Warren M. Grill, Ph.D.
Michael W. Keith, M.D.
Kevin L. Kilgore, Ph.D.
Joseph M. Mansour, Ph.D.
Wendy M. Murray, PhD.
P. Hunter Peckham, Ph.D.
David L. Wilson, Ph.D.

Cleveland FES Center

Departments of Biomedical Engineering,
Mechanical and Aerospace Engineering,
and Orthopaedics
Case Western Reserve University
and MetroHealth Medical Center

1. SYNTHESIS OF UPPER EXTREMITY FUNCTION

The overall goals of this project are (1) to measure the biomechanical properties of the neuroprosthesis user's upper extremity and incorporate those measurements into a complete model with robust predictive capability, and (2) to use the predictions of the model to improve the grasp output of the hand neuroprosthesis for individual users.

1. a. BIOMECHANICAL MODELING: PARAMETERIZATION AND VALIDATION

Purpose

In this section of the contract, we will develop methods for obtaining biomechanical data from individual persons. Individualized data will form the basis for model-assisted implementation of upper extremity FNS. Using individualized biomechanical models, specific treatment procedures will be evaluated for individuals. The person-specific parameters of interest are tendon moment arms and lines of action, passive moments, and maximum active joint moments. Passive moments will be decomposed into components arising from stiffness inherent to a joint and from passive stretching of muscle-tendon units that cross one or more joints.

Progress Report

1. a. i. MOMENT ARMS VIA MAGNETIC RESONANCE IMAGING

Abstract

A manuscript is being prepared describing our previously reported wrist moment sensor. We have begun analysis of a potential problem of aligning the axes of data measured under different conditions. We have also established a method of measuring the alignment of joint moment axes and kinematic axes.

Progress Report

In this quarter, we have identified a problem that must be addressed prior to collecting any more imaging or biomechanical (joint moment and angle) data. This problem is the alignment of axes in the two measurement systems. At present, the coordinate systems are established by different methods and criteria: with the wrist moment sensor, the axes are established by how the device is mounted on the arm; with the MR images, axes are defined by segmenting the bones, and movements are produced by the arm fixation device. Thus, there is no guarantee that the axes align, nor that movements (or static positions) are the same for the MRI and wrist moment measurements.

We are addressing this problem in two ways. First, we are devising ways of measuring the relative locations of the two coordinate systems and of ensuring that they are aligned as well as possible during measurements. Second, we will perform a mathematical analysis of the sensitivity of calculated outcomes to axis misalignment.

The relative location and orientation of the two coordinate systems will be estimated by measurements. The same MR compatible bases will be fixed to the forearm and palm during each set of measurements. These bases will contain MR detectable markers so that their location relative to bone axes can be measured from the MR data. During wrist moment measurements, infrared LEDs will be attached to the same bases, along with the wrist moment sensor structure. The locations of the IR LEDs will be tracked with an Optotrak 3D kinematic measurement system. We will also maintain (as well as possible) the same angles during the MRI measurements as during the moment measurements.

This quarter, we have also begun preparing a manuscript describing the wrist moment sensor.

Plans for next quarter

We will complete the moment sensor paper, and we will continue our analysis of the axis alignment problem.

1.a.ii. PASSIVE AND ACTIVE MOMENTS

Abstract

Previously, we reported on preliminary measurements to assess the effect of the positions of each finger on the passive properties of adjacent fingers. We found that the position of the metacarpal phalangeal (MP) joint of the long finger had a very strong influence on the passive properties recorded for the MP joint of the index finger in the single subject studied. The effect was much larger than anticipated. We have now verified this effect in four additional subjects. It appears that this effect is primarily due to the stretching of the skin between the fingers. It is important to identify this effect in order to make accurate clinical measurements. To our knowledge, this effect has not been reported previously in the literature, and textbooks explaining how to make passive range of motion measurements ignore this effect.

Purpose

The purpose of this project is to characterize the passive properties of normal and paralyzed hands. This information will be used to determine methods of improving hand grasp and hand posture in FES systems.

Progress Report

During this quarter we analyzed measurements made on five normal volunteers regarding the influence of each finger on the passive properties of adjacent digits. Specifically, we have measured the effect that the angle of the long finger MP joint has on the passive properties of the index finger MP joint. Although we expected that the long finger angle would have some effect at the extremes of it’s range of motion, we did not anticipate our initial finding that the long finger had an effect throughout it’s range of motion, and that the effect was similar in magnitude to the effect of wrist angle. We have now verified that this effect is consistent across a group of five normal subjects.

<u>Methods</u>

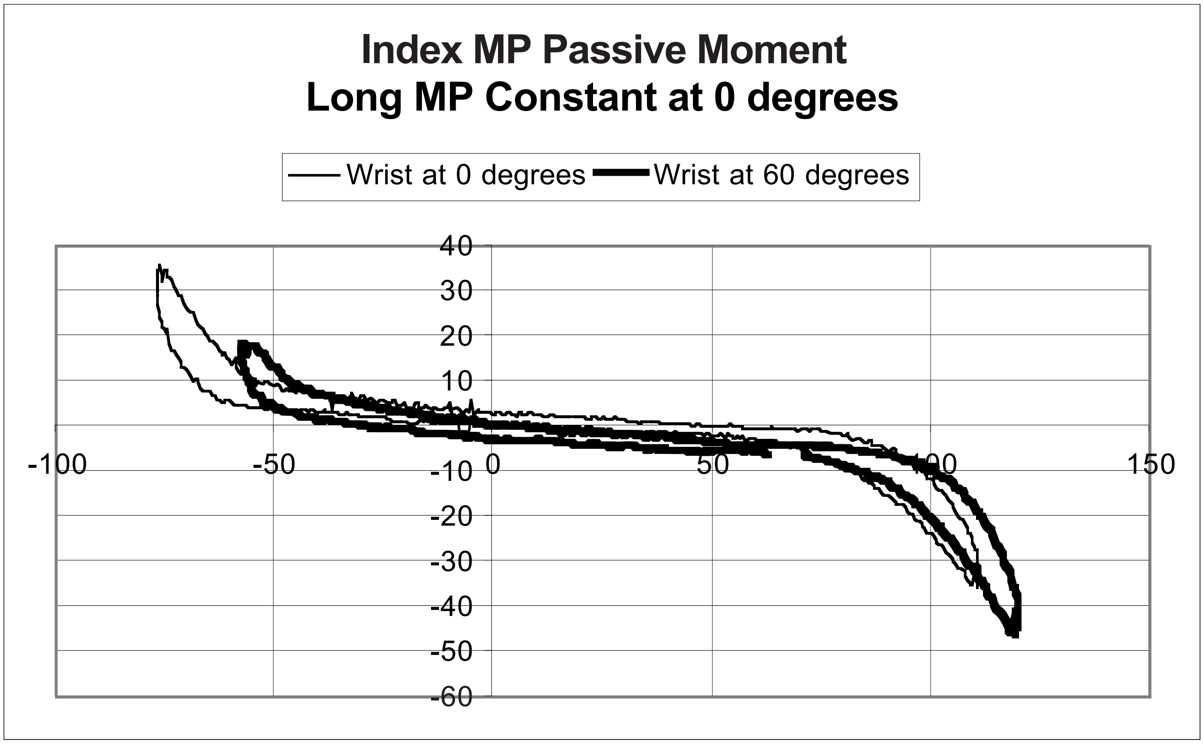
Five subjects with no prior history of joint problems were studied (age range 20 to 37). Both the index and long fingers of the left hand were splinted so that both the proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints were fixed at 0 degrees extension. The splint attached to the index finger was connected to the passive moment apparatus that has been described previously (QPR#1). This apparatus moves the finger back and forth through its entire range of motion while measuring both joint angle and joint moment. The long finger was strapped down using Velcro straps at various angles of the MP joint.

The long finger MP joint was initially fixed at 45° flexion and the passive torque-angle curve of the index MP joint was recorded using previously established protocols (QPR#2). The long finger MP joint was then positioned at 0° and 90° flexion and the index MP passive torque-angle curve recorded at each position. This complete set of measurements was made with the wrist at 0° and at 60° of flexion. At the end of the experiment, the long MP joint and the wrist joint were returned to their initial positions and the measurement repeated.

Results

In all five cases, the passive properties of the index MP joint were strongly influenced by the position of the long MP joint. The index MP extension range increased by 35° to 70° when the long MP joint was moved from 90° flexion to 0° flexion. In one subject, this change was twice as large as the change in the passive range due to a 60° change in wrist angle. In all cases, the change in extension range due to the long MP joint angle was larger than the change in the flexion range. This is probably because the long MP joint was not positioned in hyperextension for these measurements.

The results of this study for one subject are shown in Figure 1.a.ii.1. For this subject, when the long finger MP joint was fixed at 0°, there was a change in the extension range of approximately 20 degrees when the wrist angle was changed from 0° to 60° (solid lines). This is the typical effect due to the stretching of the tendons crossing the wrist joint, which we have reported previously. However, when the long finger MP joint was fixed at 90° flexion, there was no effect on the passive range due to wrist angle change. Since the effect of the tendons cannot be eliminated, we must conclude that their effect is masked by the effect of the long finger MP joint. In addition, this effect is almost certainly due to the stretching of the skin rather than to inter-tendinous connections, since we would expect that the effect of inter-tendinous connections would be altered by changes in wrist angle.



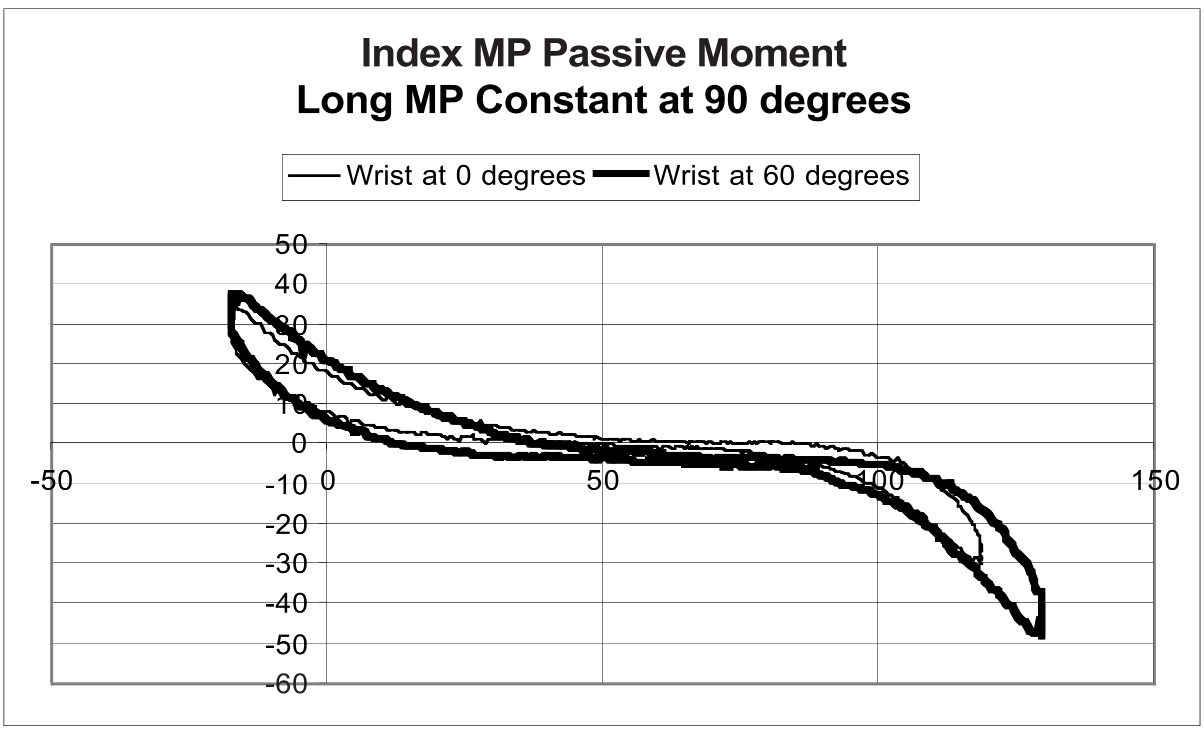


Figure 1.a.ii.1 Index MP range of motion as a function of the MP angle of the long finger and wrist angle. X-axis shows Index MP angle, Y-axis shows Index MP moment. In the top graph, the Long MP is held constant at 0 degrees, showing the typical effect of wrist angle. In the bottom graph, the Long MP is held constant at 90 degrees, where it masks any effect of wrist angle on the index passive properties.

Discussion

We anticipated that the position of the long finger would have an effect on the passive properties of the index finger. However, we expected to find that the effect was significant only when the long finger was positioned at extreme flexion or extension. Instead, the results appear to indicate that the effect is significant over the whole range of long finger positions, and this effect has the same order of magnitude as the effect of wrist angle. As far as we know, this effect has not been reported previously. In textbooks describing the measurement of passive range of motion, the position of the adjacent digits is not specified.

Adjacent digits can influence one another either directly through the skin or through linkages between the tendons and muscles (especially in the extensor digitorum communis). The effect that we have measured appears to be due directly to stretching the skin between the two digits, at least in some patients.

The primary clinical impact of this discovery is in the measurement of passive range over time. If adjacent digits are not positioned the same with each measurement, our results show that significant errors could occur. For example, in one subject, the error in the measurement of passive range could be as high as 50% if the position of the long finger varied 90°. We have observed clinically that range of motion measurements are made with the fingers in one of three positions: 1) free to rotate, 2) with the fingers fully flexed to keep them out of the way, or 3) with the fingers fully extended. Although the first method probably results in the smallest average error, it guarantees that the exact conditions of each measurement will not be repeatable from measurement to measurement. During the next quarter, we will compare these three measurement techniques in order to determine the most appropriate and clinically simple method.

Plans for Next Quarter

During the next quarter, we will perform additional experiments to examine the effect of adjacent digit position on passive properties. We will also continue the analysis of the splintless passive moment device described in the previous progress report.

1. b. BIOMECHANICAL MODELING: ANALYSIS AND IMPROVEMENT OF GRASP OUTPUT

Abstract

The ability to extend and flex the wrist provides a means to grasp and release light objects, even in the absence of voluntary hand function. Extending the wrist flexes the fingers and thumb, providing a passive hand grasp. Conversely, wrist flexion acts to extend the fingers and thumb, opening the hand. This combination of passive hand closing and opening is referred to as a tenodesis grasp. Neuroprosthesis users commonly undergo a Br-ECRB tendon transfer to provide strong voluntary wrist extension and functional electrical stimulation of the finger and thumb muscles is utilized to augment the tenodesis grasp. Voluntary wrist flexion is generally absent in individuals with high level tetraplegia, but passive hand opening can still be achieved by positioning the arm so that gravity acts to flex the wrist. This quarter we have utilized a graphics-based computer model of the upper extremity to assess how the passive wrist extension moment generated by the Br-ECRB transfer influences gravity-assisted wrist flexion.

Objective

The purpose of this project is to use the biomechanical model and the parameters measured for individual neuroprosthesis users to analyze and refine their neuroprosthetic grasp patterns.

In the past quarter, we have evaluated how the passive moment-generating capacity of the tight and slack Br-ECRB transfer (described in previous progress reports) influences gravity-assisted wrist flexion. The net passive moment at the wrist joint (before a Br-ECRB transfer) was compared to the passive wrist extension moment generated by the transfer to estimate the range of wrist postures where gravity-assisted wrist flexion is possible.

Progress Report

Simulation of Gravity-Assisted Wrist Flexion

The net passive moment at the wrist joint is determined by the gravitational moment produced by the weight of the hand and the passive properties of the wrist joint. The gravitational moment is determined by the weight of the hand and the perpendicular distance between the center of mass of the hand and the wrist joint center (i.e., the moment arm). The moment arm, and therefore, the gravitational moment, varies as a function of wrist position. Passive properties of the wrist joint are determined by joint structures, such as ligaments, and the passive moments developed by muscles that cross the wrist. Passive joint properties also vary as a function of joint position.

The net passive moment at the wrist joint was calculated by summing the gravitational moment of the weight of the hand with passive wrist joint properties measured in an individual with C5 level tetraplegia (without a Br-ECRB transfer, Lemay and Crago, 1997). The gravitational moment of the weight of the hand was calculated using regression equations (McConville *et al.,* 1980) which determined the mass of the hand and the location of the center of mass for a 50th percentile male (180 cm, 75 kg). The distance between the center of mass of the hand and the center of the wrist joint was calculated using the musculoskeletal model of the upper extremity that is described in previous progress reports. When the shoulder is abducted 90˚ and the forearm is positioned midway between pronation and supination (neutral), the net passive moment at the wrist joint is a flexion moment between 40˚ wrist extension and 26˚ wrist flexion, and an extension moment between 26˚ and 40˚ flexion (Fig. 1.b.1). At 26˚ wrist flexion, the net passive moment at the wrist joint is 0 Nm. Thus, in the absence of voluntary or stimulated muscle function, the equilibrium position of the wrist is 26˚ flexion. Importantly, an externally applied wrist flexion moment or a muscle-actuated wrist flexion moment is required to reach flexion angles greater than the equilibrium position.

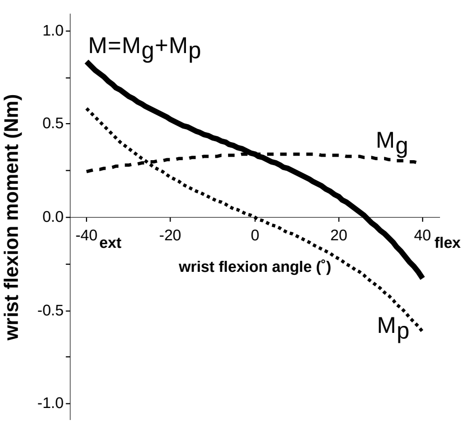


Figure 1.b.1. The net passive moment (M) at the wrist joint varies as function of wrist position and is equal to the sum of the gravitational moment (Mg) generated by the weight of the hand and passive wrist joint properties (Mp). Positive angles and moments indicate wrist flexion; negative angles and moments indicate wrist extension.

The magnitude of the passive wrist extension moment generated by the Br-ECRB transfer depends on elbow position, surgical technique, and wrist position (Fig. 1.b.2). At 90˚ elbow flexion, the slack transfer does not generate any passive moment between 40˚ extension and 40˚ flexion. Similarly, the tight transfer only generates a minimal extension moment in the most flexed wrist postures. However, when the elbow is extended, both the slack and tight transfer operate at fiber lengths long enough to generate passive force (i.e., fiber lengths greater than optimal fiber length). The passive extension moment generated by both transfers increases with wrist flexion because the muscle fibers of the Br-ECRB transfer lengthen as the wrist is flexed. The passive extension moment generated by the Br-ECRB transfer shifts the equilibrium position of the wrist toward more extended wrist postures (Fig. 1.b.3). When the elbow is fully extended, the equilibrium position of the wrist is 12˚ wrist flexion for the slack transfer and 18˚ wrist extension for the tight transfer.

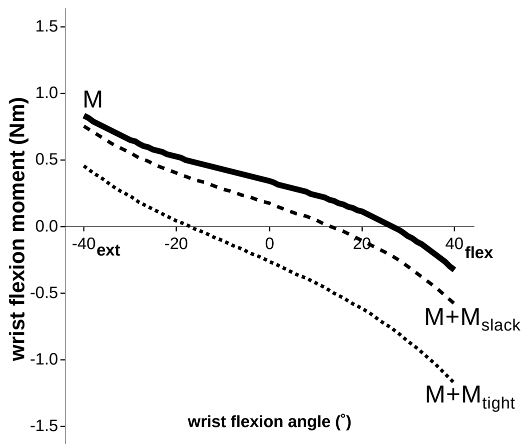


Figure 1.b.2. The passive extension moment generated by the Br-ECRB transfer depends on surgical tensioning, elbow posture, and wrist position. When the elbow is flexed 90˚, the slack transfer does not generate a passive moment, while the tight transfer generates a minimal moment in the most flexed wrist postures. In contrast, the passive extension moment generated by both the tight and the slack transfers increases with wrist flexion when the elbow is fully extended.

If voluntary wrist flexion is absent, the range of motion available at the wrist decreases when the elbow is in a posture where the Br-ECRB transfer generates a passive extension moment (Fig 1.b.4). For example, the model simulations indicate that the wrist cannot passively achieve a flexed wrist posture after a tight transfer if the elbow is fully extended. Thus, an individual who relies on the tenodesis grasp would be unable to open his hand in this elbow position. Because hand opening increases with wrist flexion angle, any limitation in the wrist range of motion could interfere with an individual's ability to grasp and release objects.

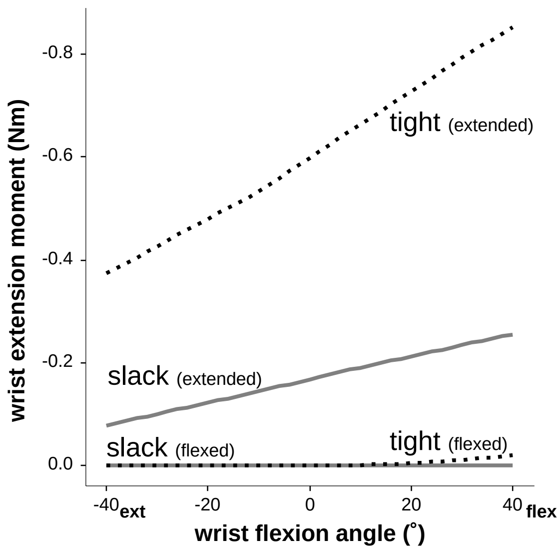


Figure 1.b.3. The net passive moment at the wrist joint decreases in arm postures where the Br-ECRB transfer generates a passive extension moment. When the elbow is extended, the equilibrium position of the wrist (the wrist flexion angle where the net joint moment is 0 Nm) shifts toward more extended wrist positions due to the passive extension moment generated by the Br-ECRB transfer.

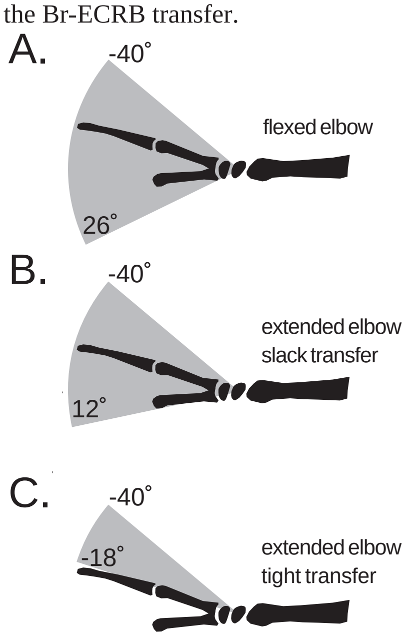


Figure 1.b.4. If the wrist is positioned between maximum extension (-40˚ flexion) and its equilibrium position, gravity will act to move the wrist to its equilibrium position unless an external or muscle-actuated wrist extension moment is applied. The shaded gray regions illustrate the wrist range of motion available in different elbow postures if active wrist flexion is absent. (A) When the elbow is flexed 90˚ the range of motion for both the tight and slack transfers is identical to the range of motion before transfer because neither transfer generates a passive extension moment in this elbow posture. (B). When the elbow is fully extended, the range of motion decreases by 14˚ due to the passive extension moment generated by the slack transfer. (C). Due to the large passive extension moment generated by the tight transfer in full elbow extension, the range of motion of the wrist is limited to 12˚ over the most extended wrist postures.

Plans for Next Quarter

In the next quarter, we plan to further investigate the effects of surgical tensioning of the Br-ECRB transfer on wrist function. The model simulations indicate that a tight Br-ECRB transfer provides better wrist extension but limits gravity-assisted wrist flexion. We plan to further investigate this trade-off in wrist function, as well as perform sensitivity studies to evaluate the robustness of our model results.

References

Lemay, M. A. and Crago, P. E. (1997) Closed-loop wrist stabilization in C4 and C5 tetraplegia. *IEEE Transactions on Rehabilitation Engineering* **5**:224-252.

2. CONTROL OF UPPER EXTREMITY FUNCTION

Our goal in the five projects in this section is to either assess the utility of or test the feasibility of enhancements to the control strategies and algorithms used presently in the CWRU hand neuroprosthesis. Specifically, we will: (1) determine whether a portable system providing sensory feedback and closed-loop control, albeit with awkward sensors, is viable and beneficial outside of the laboratory, (2) determine whether sensory feedback of grasp force or finger span benefits performance in the presence of natural visual cues, (of particular interest will be the ability of subjects to control their grasp output in the presence of trial-to-trial variations normally associated with grasping objects, and in the presence of longer-term variations such as fatigue), (3) demonstrate the viability and utility of improved command-control algorithms designed to take advantage of forthcoming availability of afferent, cortical or electromyographic signals, (4) demonstrate the feasibility of bimanual neuroprostheses, and (5) integrate the control of wrist position with hand grasp.

2. a. HOME EVALUATION OF CLOSED-LOOP CONTROL AND SENSORY FEEDBACK

Abstract

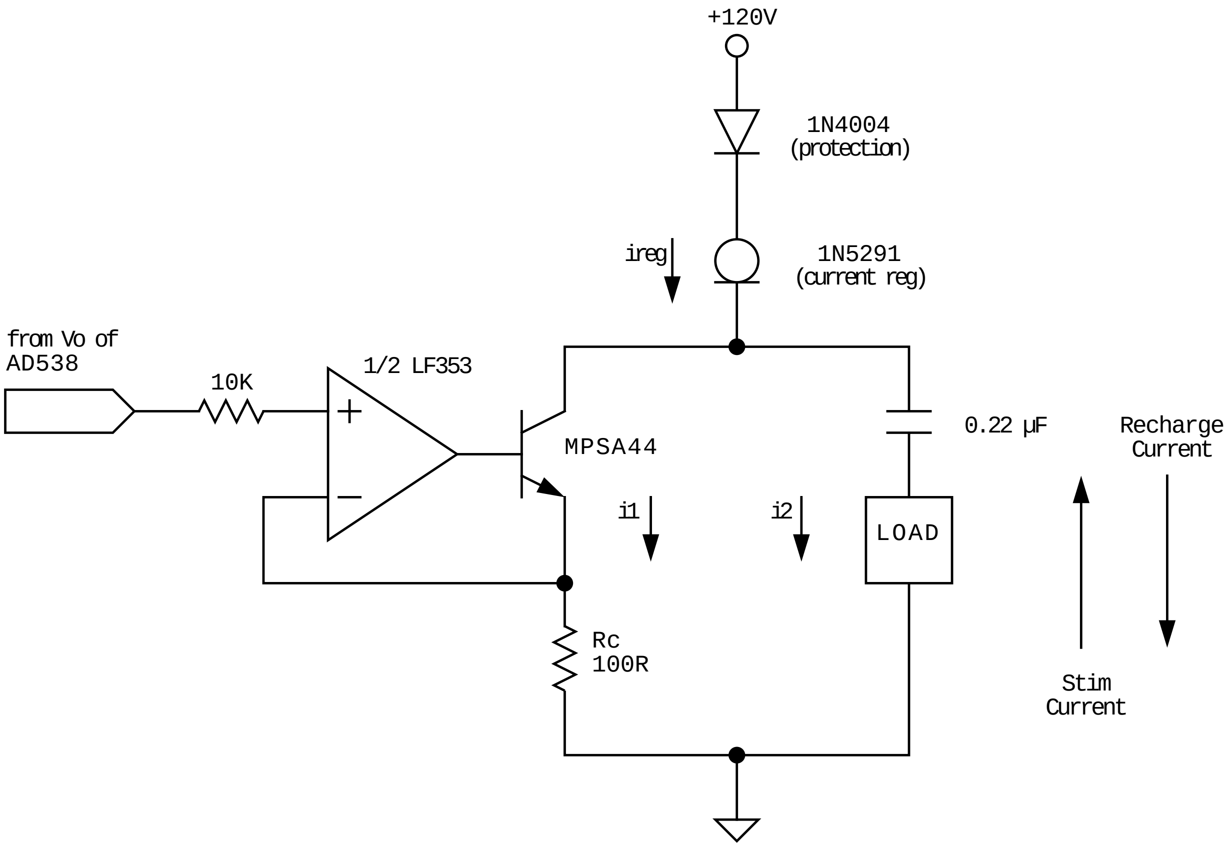
The purpose of this project is to deploy an existing portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. We have completed the development of a stand alone, analog, single channel stimulator for grasp-force feedback. A working hard-wired, battery powered prototype has also been built. The prototype was submitted to the CWRU Technical Development Laboratory for cost/time estimates for short-run production.

Purpose

The purpose of this project is to deploy a portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. Our goal is to evaluate whether the additional functions provided by this system benefit hand grasp outside of the laboratory.

Progress Report

A battery-powered, single-channel, force-feedback stimulator has been built. The completed unit measures 10 × 6.5 × 4 cm and weighs 203g (including 9V battery, sensor, and electrode cable). As described previously (9^{th} Quarterly Progress Report), it uses an FSR as a force sensor and executes a power function transformation between measured force and output pulse current using an analog devices



AD538 real-time analog computational unit. The unit is powered by a 9V battery that is used to derive ±5V and +120V supplies (the latter for the output stage only) using switched micro-power DC-DC converters (LT1111s). The major change since the previous report has been in the output stage. We had planned originally to use a modified Howland current pump in a master-slave configuration, but have opted instead for a single op-amp design used in previous stimulators built here and elsewhere (Fig. 2.a.1). The output is capacitively coupled with a recharge current limit of roughly 0.5 mA. Bench tests have shown that the stimulator is capable of generating 150 µs, 30 mA pulses across Ag/AgCl adhesive electrodes attached to the skin — well above the anticipated maximum comfortable stimulus. The total supply current drawn by the circuitry is dominated by the AD538, which draws 10 mA of the 15.5 mA consumed by the complete unit.

Our plan is to produce 5 units and distribute them to volunteer neuroprosthesis users in order to solicit their observations regarding feasibility and utility. We have submitted the design to the CWRU Technical Development Laboratory and are waiting for a cost/time estimate for production. We may also solicit bids from other, local electronic device manufacturers.

Plans for Next Quarter

We plan to initiate short-run production of the stimulators in the next quarter.

2. b. INNOVATIVE METHODS OF CONTROL AND SENSORY FEEDBACK

2. b. i. ASSESSMENT OF SENSORY FEEDBACK IN THE PRESENCE OF VISION

Abstract

The purpose of this project is to develop a method for including realistic visual information while presenting grasp-force feedback information simultaneously, and to assess the impact of force feedback on grasp performance in the presence of such visual information. In this quarter, acquire-and-hold evaluations were completed by the initial pool of 8 subjects. The data were analyzed for effects of task parameters (e.g. target window size) and the presence or absence of force feedback. We also collected additional libraries of video clips suitable for our evaluation system.

Purpose

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance. Vision may supply enough sensory information to obviate the need for supplemental proprioceptive information via electrocutaneous stimulation. Therefore, it is essential to quantify the relative contributions of both sources of information.

Progress Report

We have collected additional data from able-bodied subjects using the evaluation system and experimental protocol described previously. In brief, subjects controlled a simulated (digitized video) neuroprosthesis to complete a grasp-and-hold task; and success rates were measured as a function of the size of the target force window, with and without electrocutaneous feedback of grasp force information. Additionally, video clips were collected from two neuroprosthesis users to complete the video library.

During this quarter we performed preliminary trials on subjects and arrived at a fixed number of force target window sizes against which to evaluate all subjects. Using these standardized window sizes, we collected data from 8 subjects over 4 sessions each. Five of these subjects were presented with visual feedback (i.e., the video simulation) first and visual and force feedback second, during each session. The rest of the subjects received visual and force feedback first and visual feedback alone second. The data was analyzed using a repeated-measures, mixed-model ANOVA applied to the arcsine transformation of the success rates. The success rates ρ were subjected to an arcsine transformation [1] of the form:

$$\rho^* = 2 \arcsin\sqrt{\rho}$$

in order to ensure uniform detectability of differences in rate across the entire range of rates. The within subject (repeated) factors were window size, feedback condition, and session number. The across subject factor was order of feedback presentation.

The main result of interest is that force feedback significantly improved performance (arcsin-transformed success rates averaged across all subjects and conditions) by 14% (p=0.0041). Window size was also highly significant (p=0.0001), as expected, since making the target window narrower invariably made the task more difficult. Moreover, the improvement in performance with feedback was consistent for all window sizes (Fig. 2.b.1), as indicated by the insignificant interaction between the feedback and window size factors (p=0.1186). Initially, we expected the interaction to be significant since performance tended to saturate for larger windows, potentially diminishing the benefit of feedback. Instead, the differences between success rates with and without force feedback were equally discriminable throughout the tested range.

Fig. 2.b.2

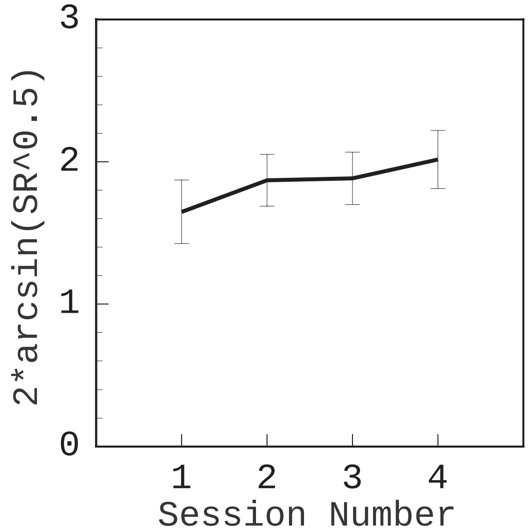
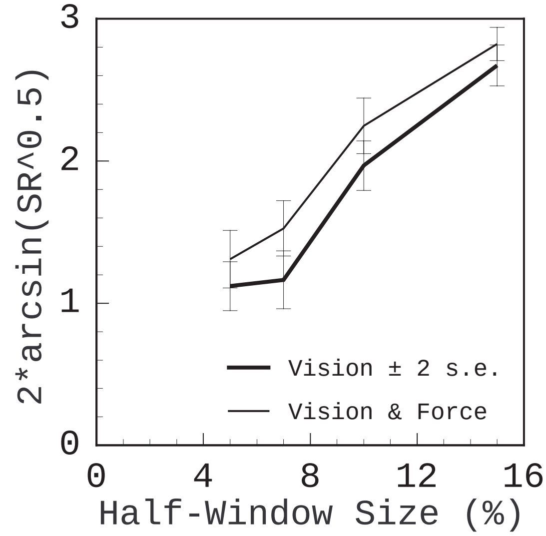


Fig. 2.b.1



Additionally, we looked at the effects of feedback presentation order and session effects. The effect of presentation order (i.e. whether a subject completed trials with vision alone or with vision and force feedback first within a session) was not significant (p=0.5910). The effect of session was significant (Fig. 2.b.2, p=0.0083), suggesting that performance improved with practice. However, the improvement apparently affected both feedback conditions equally since the interaction term was insignificant (p=0.3698), suggesting that subjects did not learn the task preferentially with feedback as opposed to without feedback.

Plans for Next Quarter

We will complete the analysis of success rates and will analyze error detection rates as well. (Recall that subjects were asked to identify the direction – too high or too low – of grasp errors in failed trials.) We anticipate submitting a manuscript describing the simulation and evaluation system and the results of the initial experiment to IEEE Transactions on Rehabilitation Engineering.

References

[1] Cohen J, 1988. Statistical Power Analysis for the Behavioral Sciences. LEA Publishers, 2nd Ed. (179-185)

2. b. ii. INNOVATIVE METHODS OF COMMAND CONTROL

Abstract

The purpose of this project is to develop new command control algorithms that will make control of neural prosthetic hand grasp simpler and more effective. During this quarter the video based evaluation system was modified to include a grasp lock routine and a new rectified-lock command control algorithm. Experimental testing was conducted to evaluate the effectiveness of the algorithm and program. The results demonstrate that the peak detection lock algorithm is superior to the normalized velocity, but the proportional controller still provided performance superior to either rectified lock algorithm.

Purpose

The purpose of this project is to improve the function of the upper extremity hand grasp neuroprosthesis by improving user command control. We are specifically interested in designing algorithms that can take advantage of promising developments in (and forthcoming availability of) alternative command signal sources such as EMG, and afferent and cortical recordings. The specific objectives are to identify and evaluate alternative sources of logical command control signals, to develop new hand grasp command control algorithms, to evaluate the performance of new command control sources and algorithms with a computer-based video simulator, and to evaluate neuroprosthesis user performance with the most promising hand grasp controllers and command control sources.

Progress Report

During the previous quarter we compared the performance of proportional control and an innovative command control algorithm called rectified lock on an acquire and hold task. The results of this evaluation indicated that the ability to lock and unlock when desired was a limiting factor in performance. Therefore, in this quarter we systematically examined two lock algorithms: peak detection and normalized lock. The prior results also indicated that attempts to unlock and re-align resulted in unwanted increases in command. Therefore, a new rectified lock algorithm with a hard threshold for increasing command was implemented. These algorithms were evaluated using the video simulator and an acquire and hold task.

1. Implementation of Lock Algorithms

Two algorithms to lock and unlock the grasp have been implemented in the video based evaluation system, peak detection (PD) and normalized velocity (NV). The PD algorithm compares the instantaneous velocity to a predetermined threshold, requires the presence of a reversal in velocity, and requires a minimum movement size. This is the algorithm used in our previous evaluations of rectified lock (see QPR 10). The NV algorithm divides the instantaneous velocity of the shoulder by the net size of the movement during a specific time period, and is the algorithm that is presently used in the hand grasp neural prosthesis. This algorithm has been implemented in the video evaluation software to compare performance to the PD lock algorithm implemented during the previous quarter.

Both the NV and PD algorithms require selection of several parameters (velocity threshold, time of peak velocity, lock movement threshold, delay time) which are used to detect the desire to lock. Since the performance of the algorithms is sensitive to the parameter selection a standardized initialization routine was implemented during this quarter (fig. C.2.b.ii.1). Five rapid shoulder elevations are made by the subject as they would do when intending to lock or unlock the hand grasp. The parameters for lock detection are determined for each trial and averaged to provide parameters for that experimental session.

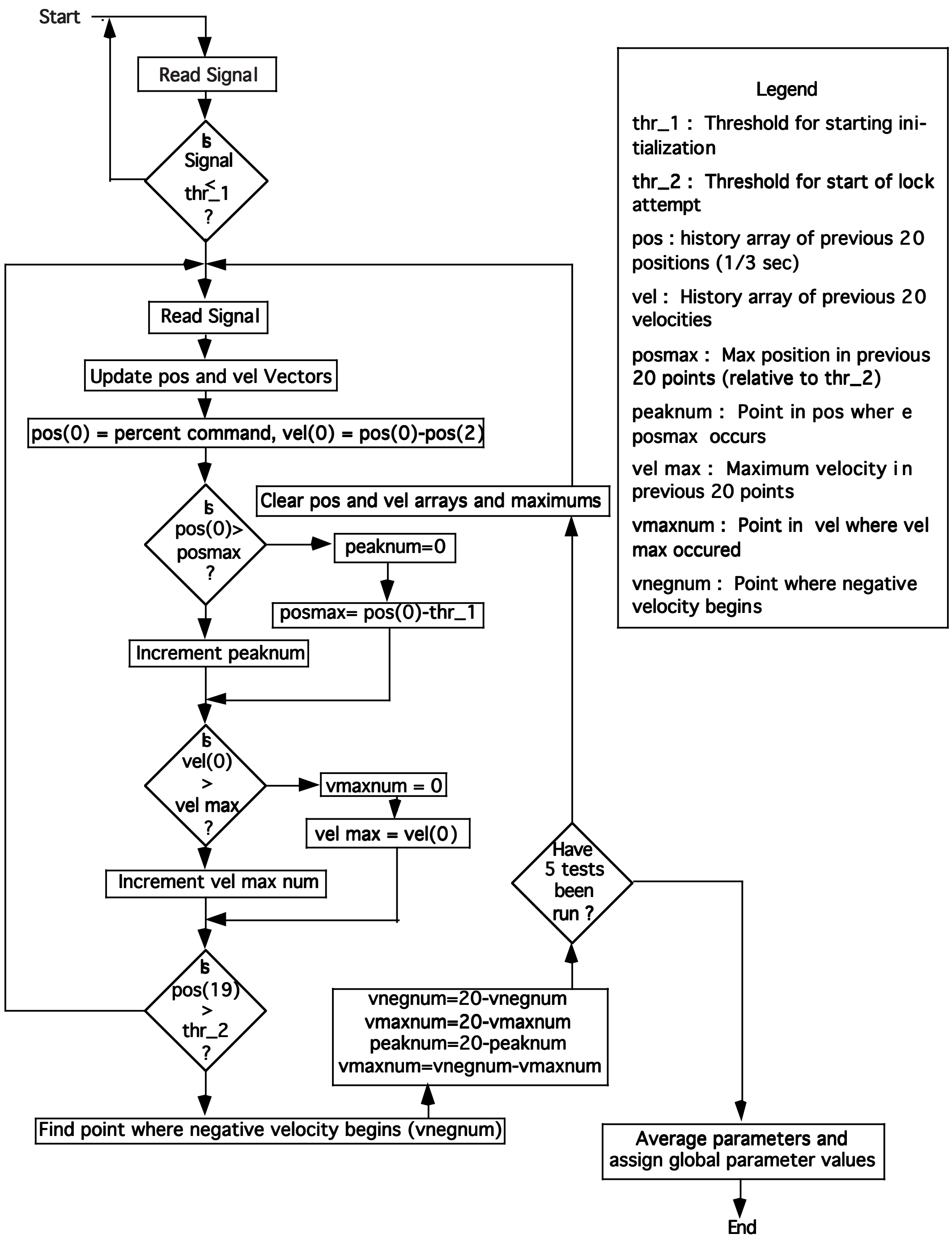


Figure C.2.b.ii.1. Block diagram of the initialization routine to determine the parameters for lock detection.

2. Implementation of Command Control Algorithms

During the previous and current quarters, eight different command control algorithms have been implemented in the video-based evaluation system. Below, summaries and state diagrams for two algorithms from which data is reported are presented: proportional control and threshold rectified-lock control.

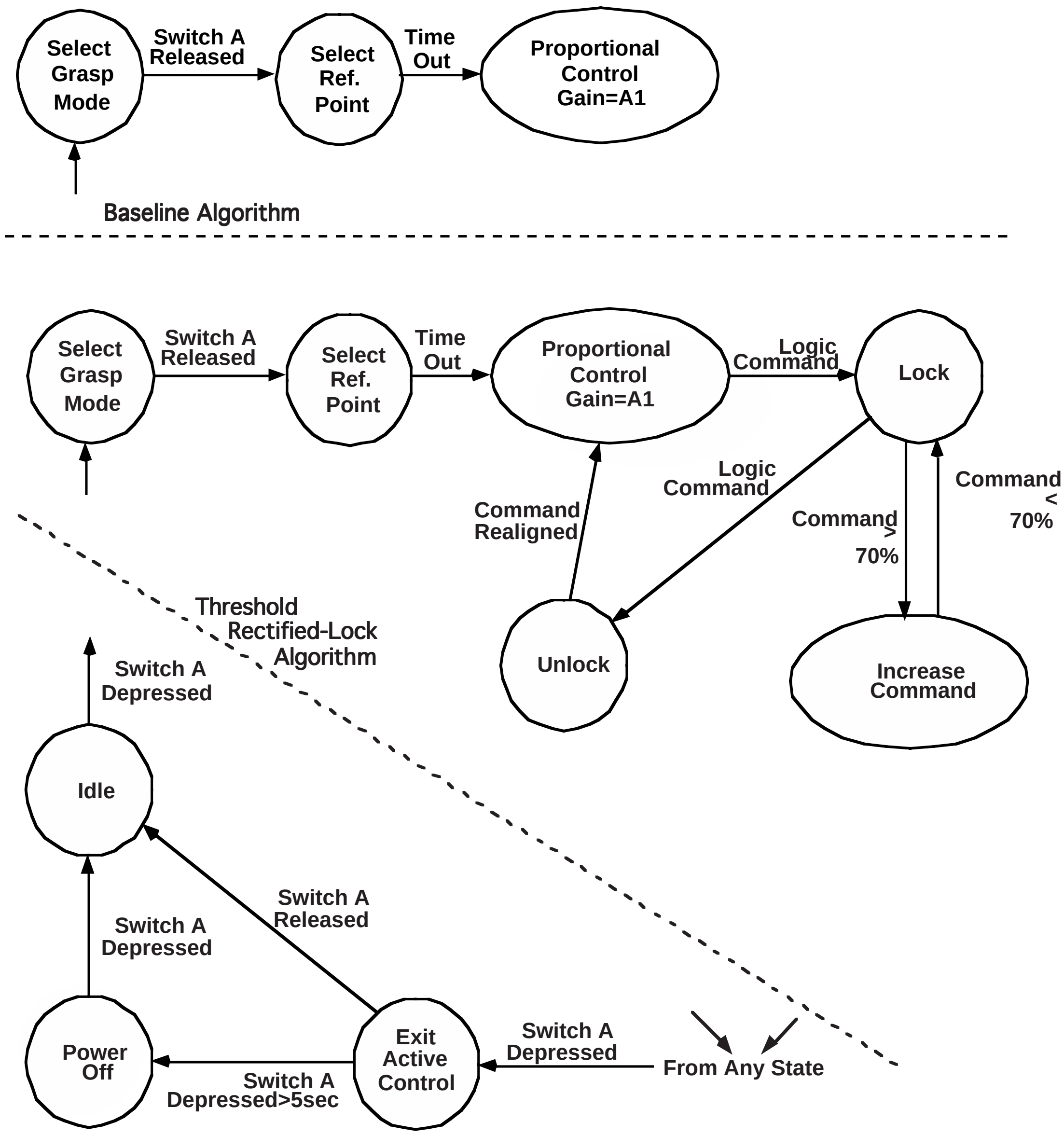


Figure C.2.b.ii.2. State diagrams of the proportional and threshold rectified-lock command control algorithms.

The baseline proportional control algorithm operates in real-time and does not use a locking routine. The state diagram for this algorithm is shown in fig. C.2.b.ii.2A. The command signal is proportional to the shoulder position and the delay between the shoulder position change and the change in command is less than 1/30 s (the video frame rate).

The threshold rectified-lock algorithm is a modified version of the proportional rectified-lock algorithm implemented and evaluated in the previous quarter (see QPR 10). These algorithms enable the users to lock the grasp at a fixed level of command (force) and then increase the command (force) with elevation of the shoulder. To increase command after lock in the threshold rectified-lock algorithm, the shoulder position must be raised beyond the 75% command position, at which time the command increases linearly with respect to time (fig. 2.C.b.ii.2B). This is in contrast to the proportional rectified-lock algorithm, in which the command increased proportionally when the command was increased above the level at which the grasp was originally locked. The rectified lock algorithm has a relatively large delay (~320 ms) as required for the lock algorithm to lock at the correct command level.

3. Evaluation of Command Control Algorithms

Testing was conducted this quarter to compare the performance of the normalized velocity (NV) and peak detection (PD) locking algorithms in combination with the threshold rectified lock algorithm. An acquire and hold task in the video-based evaluation system was used to evaluate performance. This task required the subject to use a shoulder mounted joystick transducer to generate a command signal that resulted in a grasp force within a target window within a specified time (acquire) and to maintain the grasp force within the target window for a specified time (hold). The size of the target window was modulated to alter the difficulty of the task. During our previous evaluations an acquire time to hold time ratio of 3.33 s to 6.67 s was used. In this configuration we noted that the short acquire time limited the subjects' ability to utilize the lock feature. Therefore, for one subject (txw) in the current tests an acquire time to hold time ratio of 6.67 s to 3.33 s was used. It was noted that this enabled the subject to take better advantage of the control features of the rectified-lock algorithm.

In the evaluation experiments, the subjects were first tested using the baseline proportional control algorithm, without lock, at least 2 window sizes. The number of window sizes was reduced from our previous testing to enable evaluation of more algorithms with the same number of experimental trials (prevent fatigue). The subjects next used the threshold rectified-lock algorithm with the normalized-velocity (NV) locking algorithm. Finally, the subjects used the threshold rectified-lock algorithm with the peak-detection (PD) locking algorithm. All three tests were conducted using the same window sizes.

The results, shown as percentage success in the acquire and hold task, are illustrated in fig. C.2.bii.3. Also included are results from a second testing of these 2 subjects in the same task using only proportional control. The performance of the baseline proportional control algorithm was superior to any of the rectified lock algorithms. When comparing the two locking algorithms, a consistently higher percent success was achieved using the PD algorithm than the NV algorithm. For subject *sdh* the performance between session using the proportional control algorithm was quite similar, although most points were at the saturation level of performance. In subject *txw* there was an apparent improvement in performance when comparing the 2 sessions. This may be the result of practice and the repeatability of performance across sessions will be evaluated further.

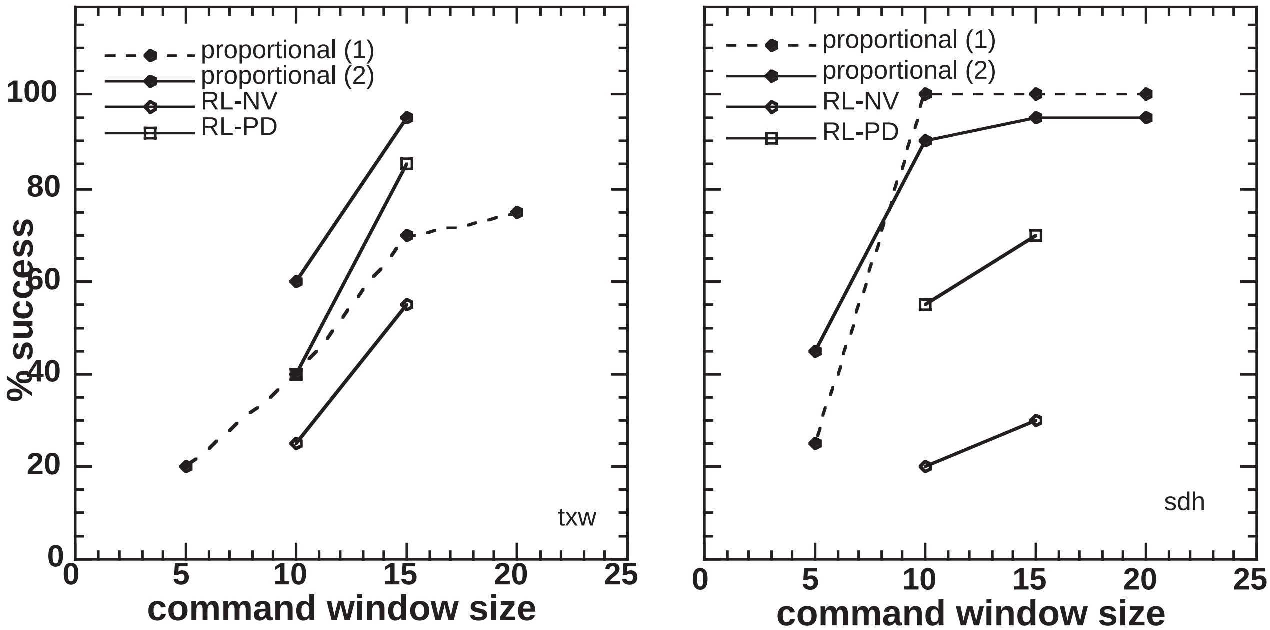


Figure C.2.b.ii.3. Performance of 2 subjects (left and right) on an acquire and hold task using either proportional control or rectified lock (RL) with lock achieved using a normalized velocity signal (NV) or by detection of the peak velocity (PD). Each point is the result of 20 trials. Sets with proportional control (1) and (2) are data collected in two different sessions.

Plans for Next Quarter

The results of testing in this quarter suggest that the simple acquire and hold evaluation task may not be sufficient to identify the performance of the modified command control algorithms. In most trials the subjects locked at a specific force level during the acquire phase and made no further adjustments during the task. In the next quarter a new evaluation task, which requires the subjects to adjust the force level, will be designed and implemented. Secondly, the large delay required for the locking algorithms, made control more difficult. A revised locking algorithm operating without the large delay will be designed and implement. Continued evaluation of command control algorithms will be conducted.

2. b. iii . INCREASING WORKSPACE AND REPERTOIRE WITH BIMANUAL HAND GRASP

Abstract

Four able-bodied subjects and one neuroprosthesis user have completed training with using the beta rhythm to operate cursor movement on a computer screen. Four of the subjects demonstrated excellent control over the beta rhythm, achieving accuracy rates above 90%. The fifth subject, while also demonstrating good control, only achieved an accuracy rate of approximately 80%. Three of the best trained subjects also participated in additional studies to address questions on EMG contamination of the signal, the ability to move and control cursor movement, and the ability to operate a neuroprosthesis with the EEG signal.

Purpose

The objective of this study is to extend the functional capabilities of the person who has sustained spinal cord injury and has tetraplegia at the C5 and C6 level by providing the ability to grasp and release with both hands. As an important functional complement, we will also provide improved finger extension in one or both hands by implantation and stimulation of the intrinsic finger muscles. Bimanual grasp is expected to provide these individuals with the ability to perform over a greater working volume, to perform more tasks more efficiently than they can with a single neuroprosthesis, and to perform tasks they cannot do at all unimanually.

Progress Report

Currently, there are five subjects (4 able bodied subjects and one neuroprosthesis user) enrolled in this study. Each subject has participated in the study for a period of at least six months. During this period of time, the subjects were trained to control the amplitude of the beta rhythm. The beta rhythm is the 18-40 Hz component of the EEG signal which is generated primarily by the frontal and somatomotor cortices [14]. The beta rhythm recorded from the frontal areas was investigated in this study since it is recorded from areas which are not directly responsible for extremity movement [15]. Secondly, this signal is not a multiple of the frequency of the electrical stimulation and thus little effect of the electrical stimulus on the recordings was anticipated.

The ability of the subjects to control the amplitude of the beta rhythm, as measured in accuracy rate, is shown in Figure 2.b.iii.1. The subjects accuracy rate is plotted as a function of time (training session). The neuroprosthesis user is given the designation NP-1, while the able bodied subjects are given the designations AB-1 through AB-4. During the six month period of time, the subjects participated in anywhere from 10 to 20 training sessions. By the end of this time, four out of the five subjects were able to achieve excellent control over the beta rhythm, achieving an accuracy rate above 90%. The achievement of the greater than a 90% accuracy rate with the beta rhythm only occurred on a consistent basis after a period of six sessions, indicating that there is a learning period involved. After this period of time, the accuracy rate is constant, indicating that once this control is learned, it is maintained. The average accuracy rate for each subject after the initial learning period is listed in Table 2.b.iii.1.

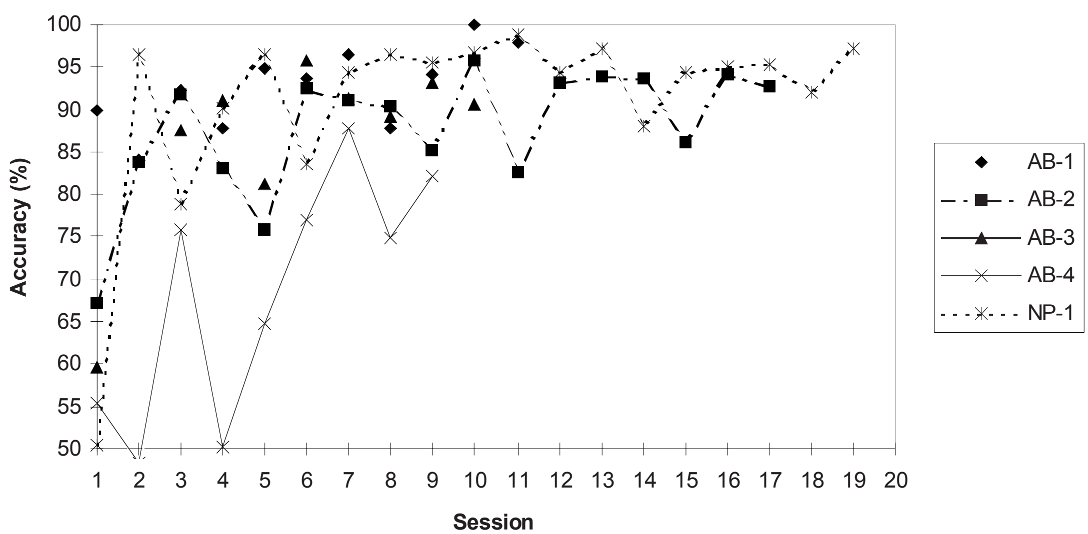
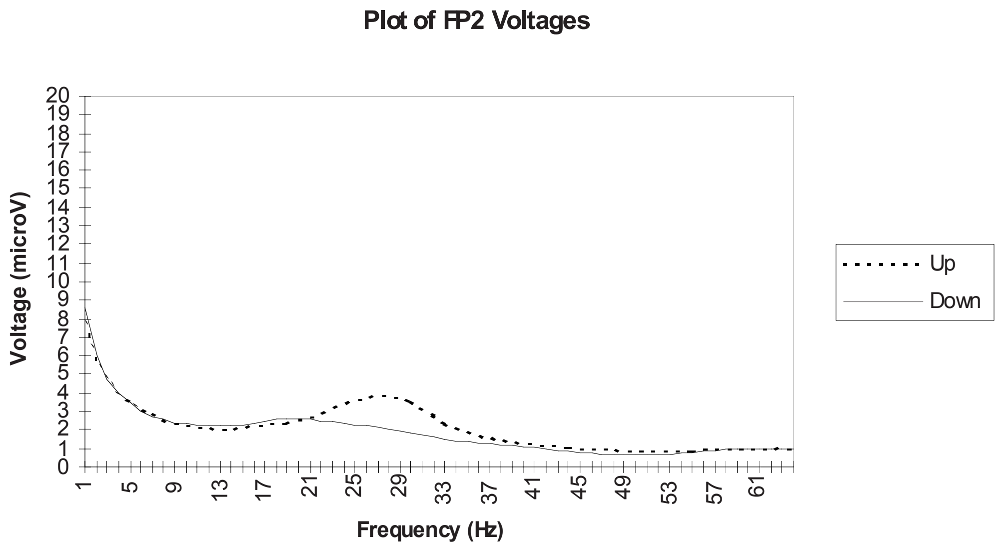


Figure 2.b.iii.1. Plot of subject accuracy over time. Accuracy rate in hitting the targets on the computer screen with the cursor is a measure of the degree of control each subject has over the beta rhythm. Subject accuracy rates reached a plateau of greater than 90% after just 6 training sessions.

	AB-1	AB-2	AB-3	AB-4	NP-1
Accuracy (%)	95.0	90.9	91.9	80.5	94.2
S.D.	4.3	4.1	2.5	5.8	4.2

Table 2.b.iii.1: Average subject accuracy rate after the initial learning period.

Subject AB-4, although achieving some control over the beta rhythm (reaching an accuracy rate of 80%), never demonstrated the same level of control as the other subjects. This is believed to be due to the erratic training schedule of this subject. This subject would participate for 1 or 2 sessions in a week, and then would not be available again for a period of 2 to 3 weeks (due to various conflicts). It is felt that it was this erratic schedule which is to blame for the low level of control. It is possible for a subject to participate in only one training session per week, or even go several weeks between sessions, only after the initial learning period (i.e. once a greater than 90% accuracy rate has been achieved for more than two sessions). If the subject is erratic in attending sessions during the learning period, full command over the beta rhythm appears not to be achieved.



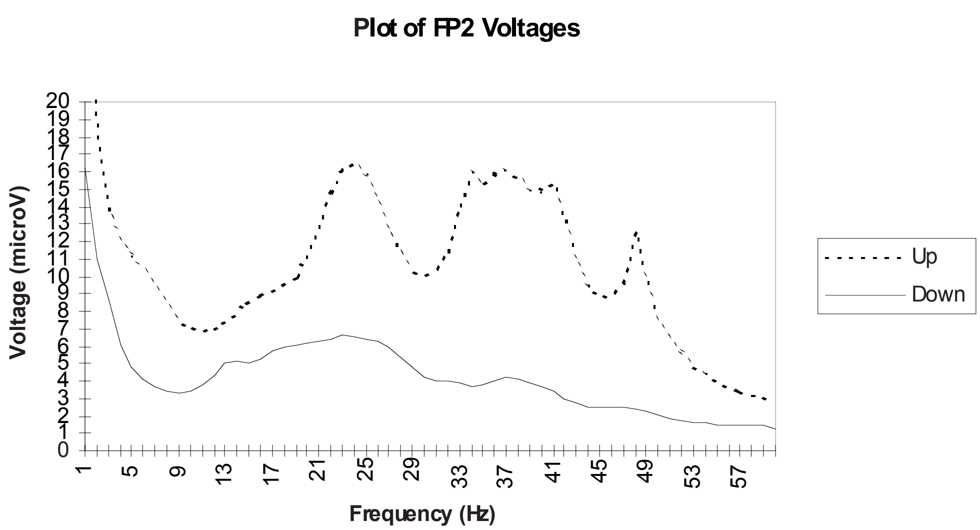


Figure 2.b.iii.2a and 2.b.iii.b: The effect of muscle activity upon the EEG signal in subject AB-2. The top figure shows what the normal separation should be for EEG control. The bottom figure shows the separation elicited by moderate muscle contraction of the forehead.

The recording of the EEG signal from the frontal areas raises concern on whether this is 'true' EEG control of whether the subject is contracting the muscles of the forehead and jaw and is using this for control of the cursor. To address this problem, three of the subjects with the greatest level of control (NP-1, AB-1, and AB-2) were asked to deliberately generate muscle activity to move the cursor. The signal generated by this was recorded and compared to the previous session in which the subject was moving the cursor as they normally would. Figures 2.b.iii.2a and 2.b.iii.2b show the spectral plots for the EEG signal for one of the subjects at the recording site which is used for cursor movement under the conditions of no muscle activity and muscle activity. As can be seen from a comparison of the figures, when there is muscle activity present, there is a greater separation of the curves, and that there is no clearly defined peak in the beta band, which occurs with training. A comparison of the EEG signal from the entire scalp shows that with muscle activity, there is wide spread activation across the entire head instead of just focused activity at only the recording site.

The three subjects which had achieved the highest level of control were also asked to participate in a study which examined the effect of upper extremity movement upon the ability to control the frontal beta. Table 2.b.iii.2 shows the effect of movement upon the subjects ability to control the beta rhythm as measured in the accuracy rate. The values given are the average accuracy rate from three runs in which a specific condition (non-movement, right side movement, or left side movement) was maintained. From this data, it can be seen that extremity movement had little effect upon the subject's ability to control the EEG signal. For Subject AB-1, there was no effect of movement upon the accuracy rate. For subject AB-2, there does appear to be some improvement in accuracy rate with movement, however, this effect was not significant. The poor results seen with right side movement by subject NP-1 were unrelated to the effects of the movement upon the EEG signal (which were not reflected in the accuracy rate with left side movement). The results seen with movement on that side are due to the fact that the cabling from the electrodes was becoming entangled with the orthosis on the right hand. It was this continual movement and pulling of the cables which resulted in the drop in accuracy rate.

Subject	**Non-movement**	**Right**	**Left**
AB-1	100 %	100%	100%
AB-2	91.5%	94.8%	96.1%
NP-1	97.9%	86.4%	96.1%

Table 2.b.iii.2: The effects of upper extremity movement upon beta rhythm control. As before, the degree of beta rhythm control is measured by the accuracy rate. For both of the able bodied subjects, there was little to no effect of extremity movement upon beta rhythm control. The poor results seen with right side movement in the neuroprosthesis user reflect complications with the recording equipment and are not an effect of movement upon beta rhythm control.

In a final study, the effects of neuroprosthesis operation upon beta rhythm control were also examined. The subject (NP-1) was asked to move the cursor while the neuroprosthesis was on and the hand was locked in a closed position. The overall accuracy rate for subject NP-1 when the neuroprosthesis was active was 93.5% (+/- 4.1% S.D.), which was only 1.5% lower than the subjects average accuracy rate without the neuroprosthesis (95.0 +/- 2.7%). However, given the large standard deviations, which exist under both conditions, the effect of the neuroprosthesis upon beta rhythm control was not significant.

The conclusions from the previous studies indicated that the frontal beta rhythm could be used as the control signal for the operation of the neuroprosthesis. To further validate this point, the BCI system developed by Wolpaw was modified so that the signal, which would normally operate cursor movement, was used to operate hand opening and closing (Figure 2.b.iii.3). The EEG signal was converted into a command signal using the gated ramp method. This method has been used previously for the conversion of the myoelectric signal [16] into a command for the neuroprosthesis. When the signal was maintained above a set threshold, this generated the command to go from hand opening to hand closing. The rate at which the hand closed was fixed (approximately 2-3 seconds to full closure), and was only generated when the amplitude of the input signal was maintained above the threshold. When the signal fell below the threshold, the hand stopped closing. To go from the hand closed to the hand open position, the signal must be below a set low threshold, which generated the command to go from hand closed to hand open at a much faster rate than hand opening (approximately 1 second to full opening).

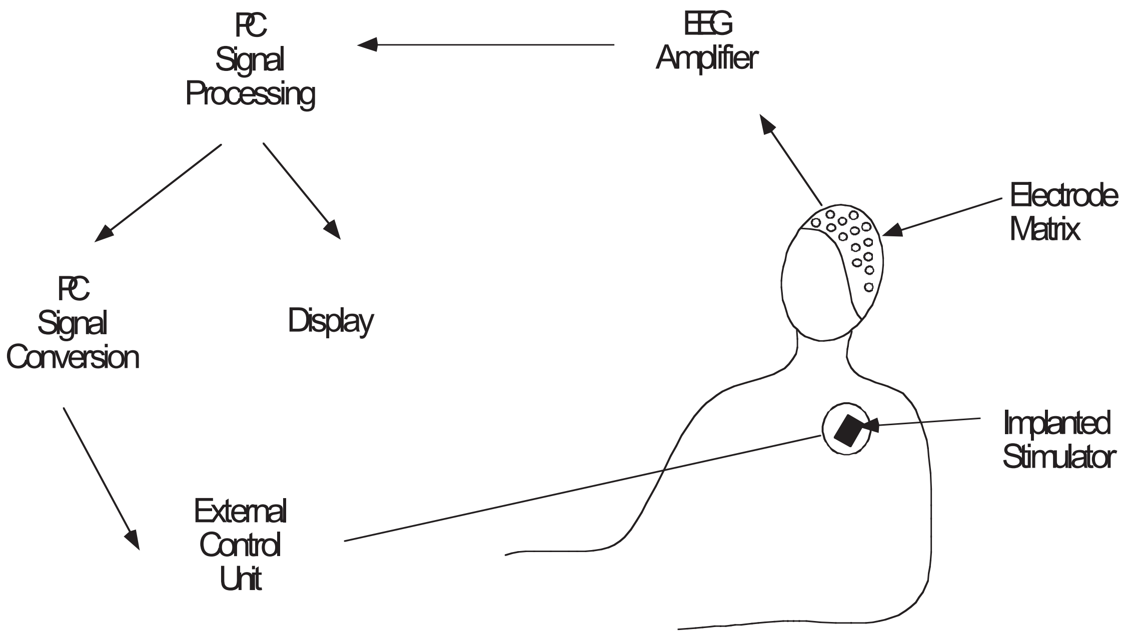


Figure 2.b.iii.3: Schematic showing the components of the EEG-based controller for the FNS hand grasp system. This system allowed for the conversion of the raw EEG signal into a command-control signal for the neuroprosthesis to provide the subject with hand opening and closing by thinking about it.

The use of the gated ramp only allowed for dynamic hand operation. Therefore, the subject was only asked to perform simple activities of daily living with the system. These tasks were pick up and move a weight, grasp and release a fork, and grasp and release a cup. The subject was also asked to open and close his hand using the EEG signal as quickly as possible, and at the verbal command of the investigator. Finally, the subject was also asked to open and close his hand in the absence of visual feedback (i.e. not looking at the hand to determine if it was open or closed).

Subject NP-1 was able to manipulate all three objects using the EEG-based controller with his neuroprosthesis. To achieve hand opening, the subject was instructed to think of moving the computer cursor up, and to think of moving the cursor down to get his hand to close. The transition between “thinking” cursor movement and generating hand opening and closing required approximately 10 minutes of training. However, there were trials where the subject was “thinking” of the wrong direction, and thus generating the wrong command to the neuroprosthesis. Another difficulty encountered was the fact that the algorithm used did not provide a means by which the subject could lock his hand in the closed position once the object was acquired. Therefore, the subject had to continually think “hand closed” while manipulating the object, which became difficult as the subject began to tire (after 1.5 to 2 hours of continual use).

Plans for Next Quarter

During the next quarter, the interface between the BCI system and the neuroprosthesis will be refined so that all signal processing and conversions will be performed on one PC computer. The algorithm to convert the EEG signal into neuroprosthetic command will also be revised to allow the user to perform hand locks and holds with minimal effort. We will also begin recruiting two more neuroprosthesis users for this study.

References

[1] M. Keith, P. Peckham, G. Thrope, K. Stroh, B. Smith, J. Buckett, K. Kilgore, and J. Jatich, "Implantable Functional Neuromuscular Stimulation in the Tetraplegic Hand," Journal of Hand Surgery, vol. 14A, pp. 524-530, 1989.

[2] M. Keith, K. Kilgore, P. Peckham, K. Stroh-Wuolle, G. Creasy, and M. Lemay, "Tendon Transfers and Functional Electrical Stimulation for Restoration of Hand Function in Spinal Cord Injury," Journal of Hand Surgery, vol. 21A, pp. 89-99, 1996.

[3] K. Kilgore, P. Peckham, M. Keith, G. Thrope, K. Stroh-Wuolle, A. Bryden, and R. Hart, "An Implanted Upper-Extremity Neuroprosthesis," Journal of Bone and Joint Surgery, vol. 79-A, pp. 533-541, 1997.

[4] Y. Handa, T. Handa, and Y. Nakatsuchi, "A voice controlled functional electrical stimulation system for the paralyzed hand," Japanese Journal of Medical Electronics and Biological Engineering, vol. 25, pp. 292-298, 1985.

[5] N. Hoshimiya, A. Naito, M. Yajima, and Y. Handa, "A multichannel FES system for the restoration of motor functions in high spinal cord injury patients: a respiration-controlled system for multijoint upper extremity," IEEE Transactions on Biomedical Engineering, vol. 36, pp. 754-760, 1989.

[6] P. Peckham, E. Marsolais, and J. Mortimer, "Restoration of Key Grip and Release in the C6 Quadriplegic Through Functional Electrical Stimulation," Journal of Hand Surgery, vol. 5, pp. 464-469, 1980.

[7] J. Chapin and M. Nicolelis, "Neural Network mechanisms of oscillatory brain states: characterization using simultaneous multi-single neuron recordings," Electroencephalography and Clinical Neurophysiology Suppl., vol. 45, pp. 113-122, 1996.

[8] D. Humphrey and J. Tanji, "What features of voluntary motor control are encoded in the neuronal discharge of different cortical motor areas," in Motor Control: Concepts and Issues, D. Humphrey and H. Fruend, Eds. New York: John Wiley & Sons, 1991, pp. 413-444.

[9] A. Schwartz, "Motor cortical activity during drawing movements: population representation during sinusoid tracing," Journal of Neurophysiology, vol. 70, pp. 28-36, 1992.

[10] G. Pfurtscheller, C. Neuper, D. Flotzinger, and M. Pregenzer, "EEG-based discrimination between imagination of right and left hand movement," Electroencephalography and Clinical Neurophysiology, vol. 103, pp. 642-651, 1997.

[11] J. Wolpaw, D. McFarland, G. Neat, and C. Forneris, "An EEG-based brain-computer interface for cursor control," Electroencephalography and Clinical Neurophysiology, vol. 78, pp. 252-259, 1991.

[12] D. McFarland, L. McCane, S. David, and J. Wolpaw, "Spatial filter selection for EEG-based communication," Electroencephalography and Clinical Neurophysiology, vol. 103, pp. 386-394, 1997.

[13] D. McFarland, L. McCane, and J. Wolpaw, "EEG-based communication and control: short-term role of feedback," IEEE Transactions on Rehabilitation Engineering, vol. 6, pp. 7-11, 1998.

[14] J. Kozelka and T. Pedley, "Beta and Mu Rhythms," Journal of Clinical Neurophysiology, vol. 7, pp. 191-207, 1990.

[15] E. Niedermeyer, "The Normal EEG of the Waking Adult," in Electroencephalography: Basic Principles, Clinical Applications and Related Fields, E. Niedermeyer and F. L. d. Silva, Eds., 2 ed. Baltimore-Munich: Urban & Schwarzenberg, 1982, pp. 71-91.
[16] P. Peckham, J. Mortimer, and E. Marsolais, "Controlled Prehension and Release in the C5 Quadriplegic Elicited by Functional Electrical Stimulation of the Paralyzed Forearm Musculature," Annals of Biomedical Engineering, vol. 8, pp. 369-388, 1980.

2. b. iv CONTROL OF HAND AND WRIST

Abstract

We have begun to specify hardware and software requirements in order to implement both a laboratory and a portable neuroprosthesis for feedforward neural network control of hand grasp and wrist angle.

Purpose

The goal of this project is to design control systems to restore independent voluntary control of wrist position and grasp force in C5 and weak C6 tetraplegic individuals. The proposed method of wrist command control is a model of how control might be achieved at other joints in the upper extremity as well. A weak but voluntarily controlled muscle (a wrist extensor in this case) will provide a command signal to control a stimulated paralyzed synergist, thus effectively amplifying the joint torque generated by the voluntarily controlled muscle. We will design control systems to compensate for interactions between wrist and hand control. These are important control issues for restoring proximal function, where there are interactions between stimulated and voluntarily controlled muscles, and multiple joints must be controlled with multijoint muscles.

Progress Report

The neural network feedforward control system that was designed and tested for simultaneous control of hand grasp and wrist position can not be implemented in the neuroprosthesis that is used clinically, and we do not have currently a laboratory system that can implement it. During this quarter, we have begun to specify the hardware and software that will be required to implement the system both in the lab, and in a portable neuroprosthesis. Our current plan is to have a laboratory system operational by September, 1999. Portable systems will probably not be available within the time frame of this contract.

The requirements of the laboratory system are the following:

1) stimulate all the hand and wrist muscles involved in either grasp mode (lateral or palmar) via either the IRS-8 or IST-10 implantable systems

2) measure via sensors, wrist flexion/extension angle, forearm orientation in the gravitational field, hand grasp opening and hand grasp force

3) compute input/output data sets from steady-state sensor data during constant stimulation with a range of stimulus parameters and combinations of parameters

4) train neural networks with the input/output data

5) implement feedforward control systems for real-time control of hand grasp and wrist angle

6) with the same sensors, measure performance during real-time control

At the present, we envision implementing a laboratory system consisting of a laboratory computer controlling an output stimulus module via a serial interface. All stimulus train control (pulse width and stimulus period on each channel) will be the responsibility of the laboratory computer. This will minimize the hardware development effort in order to start these experiments. We also envision taking maximal advantage of commercially available software (e.g. LabVIEW and MATLAB, including the neural network toolbox) to implement the control systems.

Plans for next quarter

We will continue to develop hardware and software specifications, as well as test the capabilities of the commercially available software. We expect to complete the specifications and begin testing software strategies in this quarter.